Numerical modeling of an osteoporotic femur: comparison before and after total hip prosthesis implantation

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Numerical simulation of an osteoporotic femur:
Before and after total hip arthroplasty.

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ABSTRACT: The bone strength depends on its mineralization and its geometry, which depend themselves on the supported solicitations. The bone optimizes its mass and geometry in order to improve its strength. The optimization process is called bone remodeling. This phenomenon can be deteriorated by metabolic diseases like osteoporosis. This disease weakens the bone structure and causes bone fractures. Among those fractures, femoral neck fractures (hip articulation) are the most recurrent and involve the replacement of the entire hip articulation by a mechanic articulation (Total Hip Arthroplasty).
In this paper, finite element models were developed to evaluate, firstly, the stress distribution within osteoporotic human femur bone tissue and secondly, the influence of the perturbation of the stress distribution by Total Hip Arthroplasty on its first stability. The geometry of the femur and the prosthesis was obtained by helicoid scanner acquisition. The bone was considered as two separate types of tissue: cortical bone and cancellous bone. The cortical bone was separated from the trabecular bone by apparent density threshold.
In the case of osteoporotic femur the results obtained from the simulations suggest that the mechanism of load transmission is pertubated by the bone remodelling. The degradation of trabecular architecture causes high stresses in the antero-inferior zone of the cortical bone. For the femur with hip prosthesis, the results showed that the implant has significant effects on the stress distribution within bone tissue. High stresses, due to the implant, weak the bone tissue in the lateral zone of the proximal dyaphisis and in the medial zone of the distal part at the end of the stem.
KEY-WORDS : Osteoporosis, total hip arthroplasty, 3D- reconstruction, finite element model.

1. Introduction

The osteoarticular structure holds an important place in the locomotor apparatus. It is essentially composed of bones. The structure of bone is designed to support the solicitations produced by daily activities and locomotion. In order to increase its strength and resist to daily solicitations, it adapts its mineralization and geometry by bone remodeling process. The bone remodeling can be deteriorated by metabolic diseases like osteoporosis which deteriorates bone strength. In Europe, the most consequences of osteoporosis are about 1700 bone fractures per day (W.H.O). Among these fractures, femoral neck fractures are the most recurrent and are the cause of quality of life decrease and rate of mortality increase. In general, femoral neck fracture is at the origin of the replacement of the entire hip articulation by Total Hip Arthroplasty (THA). The number of those implants tends to increase in the next decades due to the expectation of life, but also to the augmentation of the THA in youger patients.

The osteoporosis weakens the bone strength essentially by the reduction of bone mass which is due to bone structure deterioration (Hajjar et al., 2004).. The bone structure is reduced by a degradation of trabecular architecture and also by cortical envelop width decrease, and its porosity is increased (Bell et al., 1999). In the same manner, the first stability and durability of hip implants depends on bone strength which depends on its power to adapt its structure and mineralization to new stress distribution. For this, the analysis of stress distribution in bone before and after THA was a major stake and required studies on bone tissue, biomaterials as well as problems of interface between biomaterials and bone tissue.

Our aims are to study the stress distribution within human femur bone tissue and also, the influence of the perturbation of the stress distribution by hip arthroplasty on its first stability. For this, two models of the same human femur were developed. The first represents a 3D finite element model of the femur without prosthesis. The second represents the same finite element model of the femur with a cemented stem to simulate a total hip arthroplasty in the bone environment.

This work falls under multidisciplinary project in collaboration with medical and biological teams, in order to study the rupture mechanisms of bone related to strength loss, and also the orthopaedic device stability in their deteriorated environment.

2. Method

The finite element generation was performed from the geometry of a fresh human donor femur. This femur was obtained from the Laboratory of Anatomy of the Faculty of Medicine of Mediterranean University, Marseilles (responsible: P. Champsaur), and was conserved by Winkler intravenous injection and frozen at -20°C. The geometry of the femur was obtained by helicoid scanner acquisition (General Electric LightSpeed Pro 16) from the Medical imagery service of LaTimone hospital, Marseilles (0,625 mm native cuts thickness, 140 Kv). The scanner was used to generate voxel finite element model by CT2FEM (CT2FEM, 1996). This method allows to assign to each generated voxel, a density described by grey level from CT scans. Thus, each element has an effective bone density. The apparent density of the bone is calculated according to the method described by Taylor et al., 2002.

The bone was considered as two separated tissues, cortical bone and cancellous bone. the cortical bone was separated from the trabecular bone by apparent density threshold. For an apparent density greater than 0.2g/cm3, the bone was considered compact, under it was considered spongy (Bessho et al 2006). The cortical bone was considered transversely isotropic (Huiskes et al., 1981, Katz and Meunier, 1987, Pithioux, 2000) in spite of the experimental results which show a certain orthotropy.The Cancellous bone was assumed to present a large-scale isotropy (Brown and Ferguson, 1980), with strong variability according to studied region. This variation is strongly dependant on the orientation. of trabeculae.

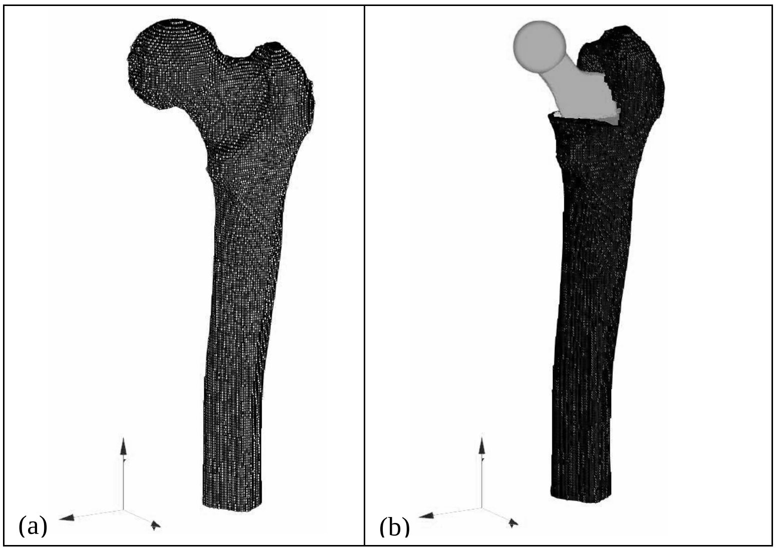

Figure 1. *(a) Maillage hexaédrique du fémur non prothésé, (b) maillage tétraédrique du fémur avec PTH cimentée.*

In this model, The compact bone elastic properties were calculated from density by a power law (Taylor et al., 2002). The grey level reported in scanners files was related to apparent density by a linear interpolation. The apparent density was then used to calculate elastic properties of cortical bone. Thus, the elastic properties of the k^{th} element were given functions of its apparent density ($\rho_{app;k}$), the maximum apparent density in femur ($\rho_{app;max}$), the maximum Young modulus in the i^{th} direction of human bone ($E_{i;max}$) and the maximum shear modulus in the ij^{th} plane of human bone ($G_{ij;max}$).

$$E_{i,k} = E_{i\,\max} \frac{\rho^2_{app,k}}{\rho^2_{app\,\max}} \; ; \; G_{ij,k} = G_{ij\,\max} \frac{\rho^2_{app,k}}{\rho^2_{app\,\max}} \qquad [1]$$

The maximum values of the elastic modulus are shown in Tab. 1. The transverse isotropy principal axis coincided with the diaphysaire axis at shaft level; with the cervical axis at femoral neck level, and with an axis in-between the two precedent axis at trochanter level. The spongy part was considered isotropic.

Elastic modulus (GPa)					
E_1	E_2	E_3	G_{21}	G_{31}	G_{32}
23	14	14	6.2	5.8	4.6

With $\nu_{23} =$ 0.42 et $\nu_{21} = \nu_{31} =$ 0.21 with the axis (1) longitudinal axis, the axis (1) et (2) radial et tangential.

Table 1. Maximal values of elastic properties in human bone (Pithioux M., 2002).

The femur model was used to generate the cemented total hip replacement by introducing an implant into femoral diaphyse. The titanium prosthesis was simulated by a homogeneous isotropic elastic solid (*E = 110000 MPa,* ν *= 0.3*). The prosthesis was fixed in bone by an acrylic cement, represented by an isotropic elastic solid (*E = 2200 MPa,* ν *= 0.3*). The interface bone-cement was supposed to be bonded while a frictional contact was assumed between cement and stem. The Coulomb's law was used with a friction coefficient of μ = 0.1 (Nuño et al., 2006)..The stem surface was considered as master and the ciment surface as slave surface.

The two models simulated one leg stance cycle of normal walking by quasi–static loading (Bergmann et al., 1993).. The load case included joint load and muscles actions. Muscles actions including abductors groups, Vastus–Lateralis and Tensor–Fascia–Lata, were applied to the anatomical attachment regions. Components of the forces are given in Tab. 2.

		Fx (N)	Fy (N)	Fz (N)
Joint load		338	-208	-1462.5
Muscle forces				
Tensor fascia Point P1	Distal	4.18	-5.85	-158.84
	Proximal	-60.19	96.97	110.35
Vastus Lateralis Point P2		7.524	154.66	-776.64
Abductors Point P2		-484.8	35.948	723.14

Table 2. Joint load and muscles forces (Newton) during one leg stance of gait cycle (Bergmann et al 2002).

The finite element model of the femur was meshed with 85624 hexahedral 8 nodes elements corresponding to three parts (the neck, trochanters, and shaft), each of them was composed of 28 parts corresponding to material groups with same elastic properties. The two bounded part of the prosthetic femur model were meshed with 395073 tetrahedral elements. The metallic stem was meshed with 64833 tetrahedral elements. The joint load was applied on femoral and implant head at the most upper node of the femoral head prosthesis according to Yoshida et al. 2002. The ABAQUS (Hibbitt, Karlsson and Sorensen, Inc.) was used to solve the problem and analyzed the stress distribution in the femur and THA.

3. Results

Frontal cut view of Von Mises stresses distribution in the cortical and cancellous regions of whole femur and THA are shown in are Fig. 2. For the whole femur, the load applied to the femoral head was transmitted from trabeculae network to the inferior face of femoral neck. In this region, the Von Mises stress ranged between *22 MPa* and *26 MPa*. In the cancellous bone of the femoral neck, the stress is less important than in the cortical one. (*~10 MPa*).

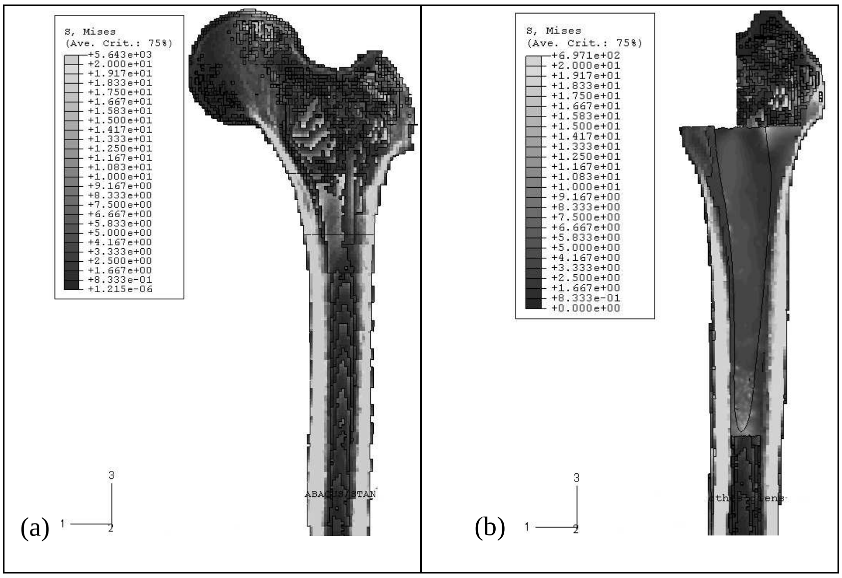

Figure 2. Frontal cut view of the *Von Mises stress distribution (a) in osteoporotic femur (b) in THA.*

For the THA model, the Von Mises stress in the medial side of the diaphysis varying from 5 MPa under the neck of the stem to 28 MPa at the end of the stem. In the lateral side, the Von Mises stress varying from *24 MPa* in the trochanter region to *7 MPa* at the end of the stem. In the ciment, the Von Mises stress varying from *8 MPa* at the top of cement to *13 MPa* at the end of the stem.

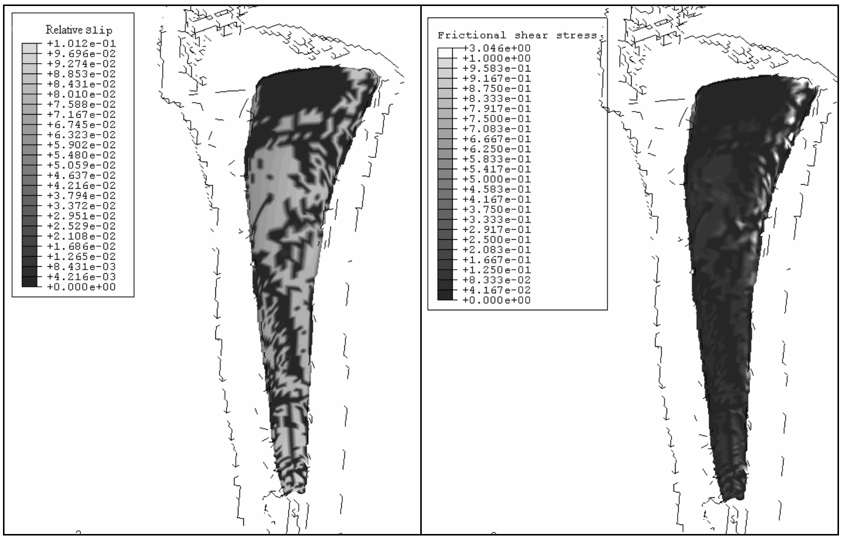

Figure 3. Frontal cut view of the relative slip between stem and ciment (a) and frictional shear stress (b) *in THA model.*

The contact formulation in the THA model allowed as estimating the micro–slip, frictional shear, and contact stress, between the stem and the cement (see figure below). The stress was concentred in medial and lateral face of the contact surface. Ranged between *4 MPa* and *7 MPa*, and was highest at proximal lateral region. The relative slip corresponding to this level of stress was ranged between *80 µm* and *100 µm.*

4. Conclusion

In this study two finite model were developed in order to study the stress distribution in human femur before and after hip replacement. The first model, represents a voxel based finite element model of human femur generated from medical imaging. This model relates bone mineralization to elastic properties with a power law (Taylor et al., 2002) in order to simulate the transverse isotropy behaviour of human bone. The second model used the same femur in order to simulate, the first stability of a cemented prothesis in contact with bone, after total hip arthroplasty.

The stress distribution calculated by the first model showed an overloading of the inferior region of cortical neck. This is probably due to osteoporosis degradation of trabecular network which ensures the load transfer and repartition over all the cortical envelop of the femoral neck. Beck et al 1999., found that femoral neck was subject to micro–structural degradation of compact tissue by essentially increasing bone porosity and reducing shell thickness. The combination of cortical bone degradation and it overloading can explain the femoral neck fractures due to osteoporosis.

In the same manner, the second model showed that the hip replacement perturbs stress distribution and involves overloading of the great trochanter region and the distal end of the stem, as soon as, underloading of the medial proximal region introduce bone mass lose and reduce the stem stability. The same observations are mentioned by Rakotomanana et al., Furthermore, in contact surface between stem and cement, the micromotion predicted was comparable to those found by Raminaraka et al., with a greater friction coefficient (0.4). Rakotomanana et al., found a micromotion value of 50 µm with a friction coefficient of 0.6.

Finaly, these models studied only the first stability without take in consideration the bone adaptation. For this, in the next step, bone remodelling process will be include in our model in order to study the long term stability and osteo–integration of orthopaedic devices in their environment.

5. Acknowledgements

The authours want to thank professor P. Champsaur and Doctor T. LeCorroller (Anatomy laboratory of medecin faculty, Mediterranean university) for having provided the bones ans realized tomography acquisitions. We also thank S. Seguinel for her assistance in the drafting of this poster.

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